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A Digital Signal Processor for Doppler Radar Sensing of Vital Signs

Editor's Note: This article tied for third-place in the 2001 EMBS Student Paper Competition. Except for minor editing and formatting changes, the article appears as it was submitted for the competition.

The treatment of thorax-related illnesses is becoming an increasing problem worldwide. An estimated 100 million Americans suffer from chronic health conditions including heart disease, lung disorders, and diabetes, and treatment for these conditions accounts for three-fourths of total U.S. healthcare costs [1]. Consequently, there is a growing market for appliances that allow both remote monitoring of health parameters and transfer of the recorded data to a physician, for convenience and cost reduction. Noninvasive sensing of circulatory and respiratory movements with a microwave Doppler radar [2] can be used for such remote monitoring. By applying telecommunication devices and frequencies for this remote sensing, the existing telecommunication network can eventually be used to facilitate the transfer of patient data to healthcare professionals [3].

Microwave Doppler radar monitoring of respiratory [4], cardiac [5], and arterial [6] movements was demonstrated with commercially available waveguide X-band Doppler transceivers in the late 1970s. In the late 1980s, a portable microprocessor-based noninvasive cardiopulmonary heart rate monitor was demonstrated [7]. Although it employed simple digital algorithms to determine the heart and respiration rates, the heart rate monitor relied on analog filters to separate the heart and breathing signals. A bank of batteries provided the power for the entire system, making portability difficult.

Because of the recent rapid expansion of wireless communications and information technology, smaller, lighter, and less expensive circuitry is readily available today. Similar advances in the field of digital signal processing (DSP) open up countless new possibilities. The speed doubling of chips every 18 months (Gordon Moore's "law") is allowing more complex calculations to be performed in ever shorter amounts of time. Bulky analog filters can now be replaced by software-implemented algorithms. These technological advances make the construction of a mobile remote monitoring station very feasible. Systems employing digital signal processing already exist [8], but these systems rely on contact measurements. While such noninvasive measurement systems do not compromise the physiological integrity of the subject, in some situations the use of contact sensors may still be disruptive or otherwise unsuitable [2]. A less disruptive noncontact system can be realized by combining a small microwave radio for Doppler radar sensing [9] with the corresponding digital signal processing software. A consequence of this system, however, is that the received signals are corrupted with much

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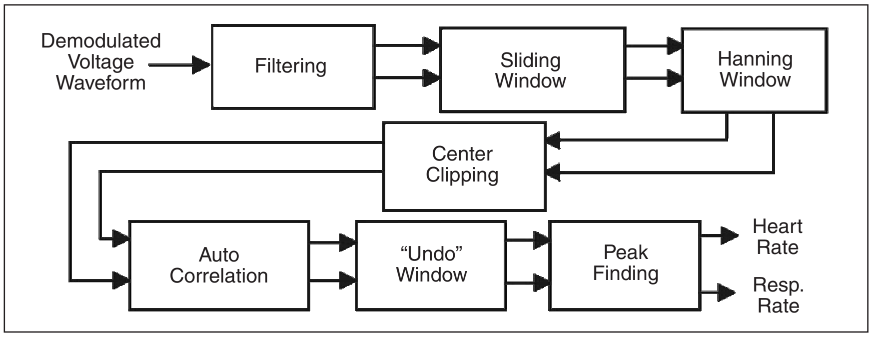
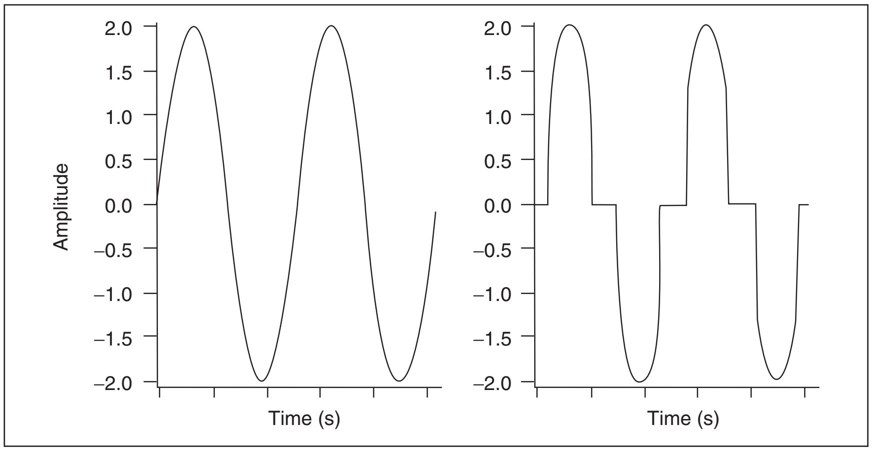


1. Block diagram of TeleSensor.



2. Illustration of the process of center clipping. *K* = 0.6, which gives a threshold of 1.2.

more noise than those obtained with contact measurements. Therefore, a robust digital signal processor is necessary to correctly extract useful data.

In this paper, a digital signal processor for Doppler radar sensing of vital signs, implemented in LabVIEW [10], is described. Several implemented techniques are described that make the processing more robust. Respiration and heart rates are successfully extracted from signals received from distances of up to 2 m from the subject.

Methodology

According to Doppler theory, a constant frequency signal reflected off an object with a periodically varying displacement will result in a reflected signal at the same frequency but with a time varying phase, $\phi(t)$. The reflected signal is effectively phase modulated (PM). If the change in displacement is small compared to the wavelength of the signal, the phase change will be small, and the PM signal can be directly demodulated by mixing it with a portion of the original signal [9]:

$$\sin(\omega t + \phi(t)) \cdot \cos(\omega t) = \sin(\phi(t)) + \text{high frequency terms} \quad (1)$$

where $\cos(\omega t)$ is the original signal with constant frequency ω, and $\sin(\omega t + \phi(t))$ the reflected signal with phase shift. The right-hand side of (1) can then be filtered for the phase shift term $\sin(\phi(t))$ [3]. The phase can be extracted from this term by using a small signal approximation: $\sin(\phi(t)) \approx \phi(t)$ for small $\phi(t)$.

Analogous to the phase shift on a transmission line terminated with a load at a varying position, this time-varying phase is proportional to the displacement $x(t)$:

$$\phi(t) = \frac{4\pi}{\lambda} x(t) \quad (2)$$

where λ is the wavelength of the signal. The demodulated signal is thus proportional to the periodic displacement of the reflecting object, as in (2). If the change in displacement is small compared to the wavelength, the demodulated signal is proportional to the periodic displacement of the reflecting object. If this object is a person's chest, the demodulated voltage waveform represents displacement due to respiration and heart activity.

A block diagram of the signal processor is displayed in Figure 1. The incoming demodulated voltage waveform was filtered in such a way as to separate the two components. The subject's normal (resting) respiration lay somewhere in the order of 20 breaths per minute. The chosen range for the respiration was between 0 and 40 breaths per minute. A fourth-order low-pass Butterworth filter with cutoff frequency at 0.7 Hz selected signals in the mentioned range. The resting heart rate of the subject varied anywhere between 60 and 120 beats per minute. A fourth-order bandpass Butterworth filter with cutoff frequencies at 1 and 3 Hz selected an adequate range of 60-180 beats per minute.

A sliding window was then passed over the waveform. All calculations were performed on the data inside the window. A typical window size was 10 s: this interval provided enough data for reliable calculation but could still track rapid changes in heart or respiration rate. The sliding window was shifted over the waveform in one-sample increments. Up to 50 samples per second could still be handled in real-time in a 10 s window. After windowing, some enhancement techniques were applied to improve the quality of the two separated signals. One technique was using a Hanning window, which prevented spectral leakage and improved the analysis of acquired signals. Unfortunately, it also left a distinct shape on the autocorrelation function response. This was remedied, however, by multiplying the result of the autocorrelation by the inverse of the Hanning window (a so called "undo" window).

Another enhancement technique used was a center clipper [11]. This function, commonly used in the processing of audio data, was used to remove unwanted peaks in the signal. The center clipper function was defined as follows:

$$\begin{aligned} c(n) &= 0 \qquad if|s(n)| \le k \cdot a_{\max} \\ &= s(n) \quad if|s(n)| > k \cdot a_{\max} \end{aligned} \quad (3)$$

where $c(n)$ was the output signal, $s(n)$ the input signal, and $a_{\max}$ was the maximum amplitude of the signal in the specified window. The user could set the factor k, and it determined the threshold at which the signal was cut off. An illustration of the process of center clipping is shown in Figure 2.

After these enhancements, the actual determination of the respiration and heart rate starts. A commonly used method to determine the period of a signal is the autocorrelation function [8]. One of the properties of the autocorrelation function is that if the input signal contains a periodic component, the autocorrelation function will contain a periodic component with the same frequency.

The resulting output signal after autocorrelation contained peaks at integer intervals of the period of the signal. The peak finding used an algorithm that fitted a quadratic polynomial to a sequential group of three samples. This group was moved in one sample increments inside the window. The slope of the second derivative of the fit was checked to determine if a peak or valley occurred. Using these peaks, the period of a given signal could be determined. The average period, taken over the first three peaks on either side of the zero-lag peak, was used. It was then straightforward to determine the rate per minute of the given signals.

The test signals used as input were taken in an anechoic chamber, using a directional antenna facing a test subject seated at a distance varying between 1 and 2 m [9]. The received signal was passed through a 12 dB/octave roll-off bandpass filter (Stanford Research Systems SR560) with 0.03-10 Hz bandwidth. This was done as a simple means to eliminate dc offset, avoid aliasing, and minimize out-of-band noise. As a result, the resolution of vital sign fluctuations in each sample was optimized. Analog-to-digital conversion was done with a HP Infinium digital oscilloscope (HP 54815A), with a sampling rate varying between 25 and 50 samples per second.

For these test signals, the success ratio was calculated. This ratio was defined as the percentage of time the calculated rate was within 2% of the reference rate. The success ratio was not calculated for the first three seconds, since this interval did not contain enough information to calculate the heart rate. Because there was no reference available for a breathing signal, it was not possible to calculate its success ratio.

Results

Some representative results of the digital signal processor are displayed in Table 1. Dataset 1 was a 20 s sample taken with a 2.4 GHz carrier wave from an antenna placed at 1 m from the subject. Dataset 2 was a 20 s sample using an 850 MHz carrier wave from a distance of 1 m. The last set, dataset 3, utilized a 2.4 GHz wave to obtain a 25 s sample also from a distance of 1 m. The first two sets had a sample rate of 50 samples per second. The last set had a sample rate of 25 samples per second. The window size used on each of the datasets was 10 s. The center clipper factor was set to 0.2, which was slightly lower than the value typically used in audio applications (0.3).

Discussion

Part of a given dataset, taken with an antenna frequency of 2.4 GHz at a distance of 1 m for a duration of 20 s, is shown in Figure 3. It can be seen that the respiration (second trace) and heart signature (third trace) can be recovered from the original raw (wide-band) signal (top trace). A finger-pressure pulse sensor (UFI-1010 pulse transducer) is used during the measurements to provide a reference signal for heart activity (bottom trace). In Figure 4, the calculated autocorrelation function is shown. From the peaks, marked with small squares, the respiration and heart rate are calculated to be 17 breaths per minute and 77 beats per minute, respectively. The calculated heart rate corresponds quite well to the reference heart rate, which is also 77 beats per minute.

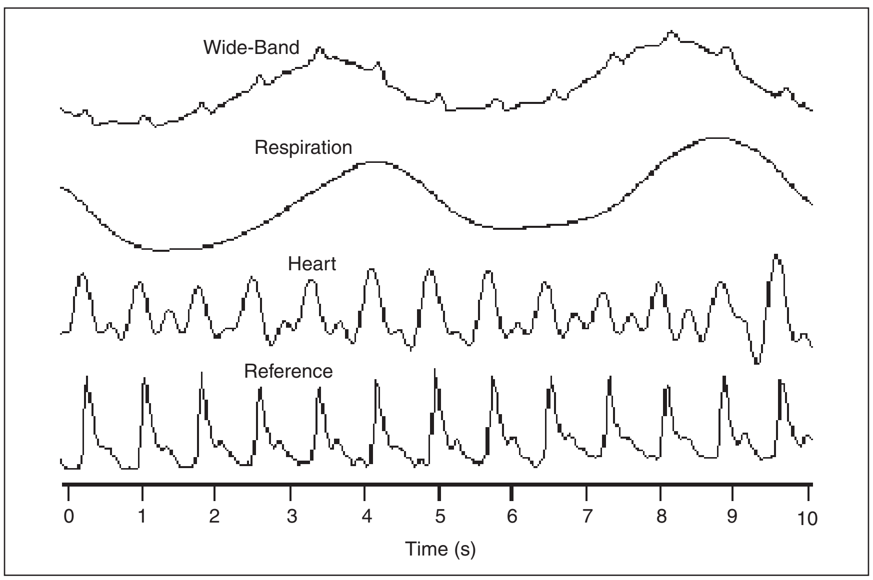
Table 1. Results of the TeleSensor.

	Success Ratio* of Heart Rate Recovery (%)
Dataset 1	93.84
Dataset 2	88.93
Dataset 2	92.53

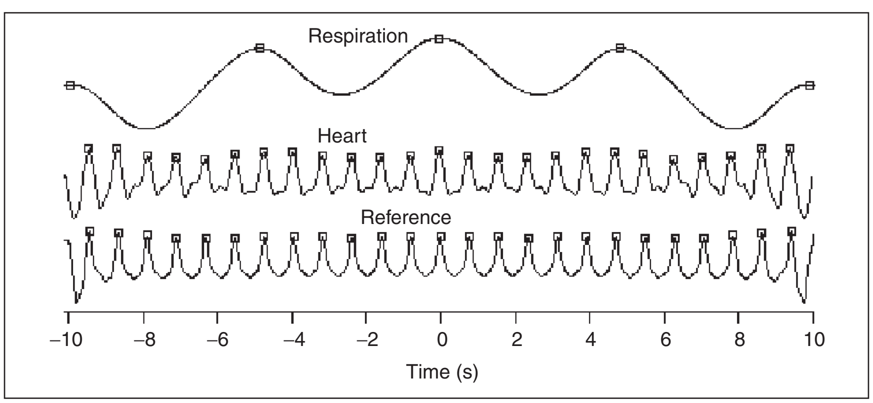
* Success ratio is the percentage of time the calculated rate is within 2% of the reference rate.

The calculated history plot of dataset 2 is shown in Figure 5. The calculated heart rate, after an initial settling time of 3 s, tracks the reference rate within a couple of beats [Figure 5(a)]. The breathing takes a bit longer to settle, about 5 s [Figure 5(b)]. This is due to the larger period of the breathing signal.

The use of the center clipper, as illustrated in Figure 6, eliminates peaks that are not associated with the fundamentals



3. Wideband (at 1 meter distance), respiration (0.00 - 0.70 Hz), heart (1.00 - 3.00 Hz), and pressure pulse reference signals.



4. Autocorrelated signals corresponding to the respiration, heart, and reference signals shown in Figure 3.

of the heartbeat. A closer analysis of the cardiac pulse, shown as "heart" in Figure 3, shows a small spike halfway through each heartbeat. This is known as the dicrotic notch, which signifies a sudden drop in pressure after systolic contraction. It is caused by a small reflux flow of blood back into the aortic valve and coronary vessels. This dicrotic notch in the heart signal is clipped. This can be seen clearly in Figure 6 at time intervals 2 and 9 s.

There are several reasons why the signal processing is done digitally. Before any information can be extracted from the demodulated voltage waveform, the heart and respiration signals need to be separated. This can be done by using hardware [7], [9], such as analog filters and amplifiers, or by using digital signal processing software. Digital processing offers implementation flexibility, filters with closer tolerances, utilizes fewer components, and has an overall lower price. DSP software not only replaces hardware filters for separation of heart and breathing signals but also provides convenient means of extracting heart and breathing rates. For these reasons, a digital signal processor called TeleSensor has been developed.

Conclusions

A signal processor for the determination of respiration and heart rates in Doppler radar measurements has been described. The processor can reliably calculate both rates for a subject at distances as large as 2 m. The rate determination is based on autocorrelation and uses several enhancement techniques, including a center clipper. Several representative results are included to show the future potential of using the processor for this purpose. Calculated heart rates agree for over 88% of the time with the reference rate, within a 2% margin, for all datasets. These results indicate excellent prospects for remote monitoring of vital signs through noncontact radar techniques.

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Bram Lohman was born in Soest, The Netherlands, on 17 February 1977. Living abroad for most of his childhood, he returned to the Netherlands for his M.Sc. degree in electrical engineering, which he received in 2001 from the Delft University of Technology. His graduation thesis, titled "Exploration and realization of a (real-time) conferencing system using the Philips 3D-LCD," concerned the transfer and modification of real-time video data for a 3-D monitor. Currently, he is taking a year off to travel around South East Asia.

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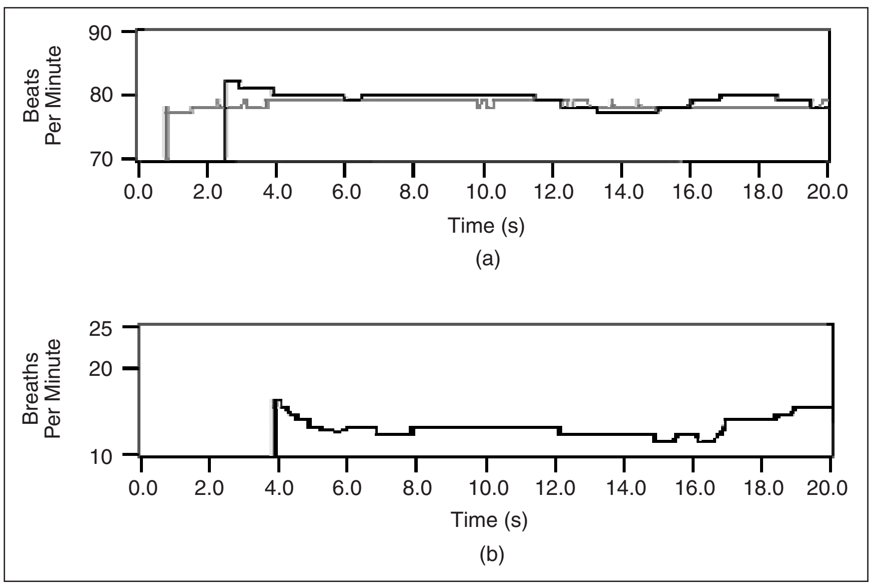
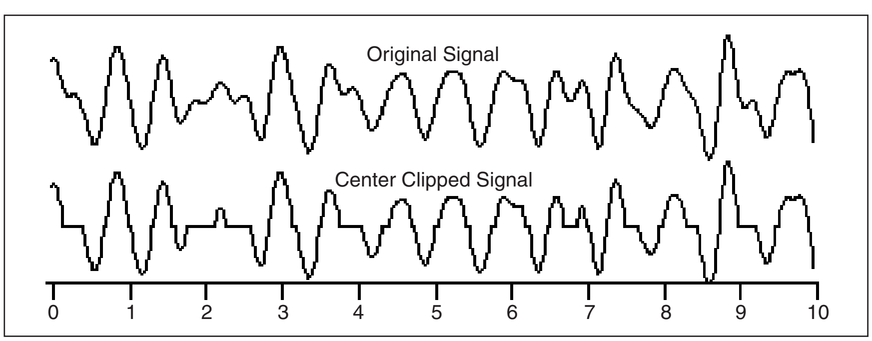


5. History windows showing the calculated (black) and reference (gray) heart rate (a) and the calculated respiration rate (b).



6. Center-clipped heart rate signal, showing the clipping of the secondary beat.

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